Ultra-low pressure sensor for neonatal resuscitator

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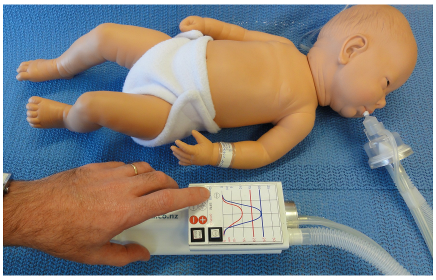
Abstract

A Venturi-type flow sensor has been designed and fabricated for neonatal respiratory assistance to control airway pressure and tidal volume. As the low flow range and sensing principle require the measurement of correspondingly very low pressures, a very responsive sensor, based on a polymer membrane acting onto a piezoresistive cantilever force sensor based on low-temperature co-fired ceramic (LTCC), was developed. This paper details the 3D modelling, manufacture, assembly and characterisation of the sensor. Compared to expensive and fragile MEMS-based devices, this sensor, based on LTCC, thick-film technology and polymer parts, provides an accurate and robust, yet low-cost alternative.

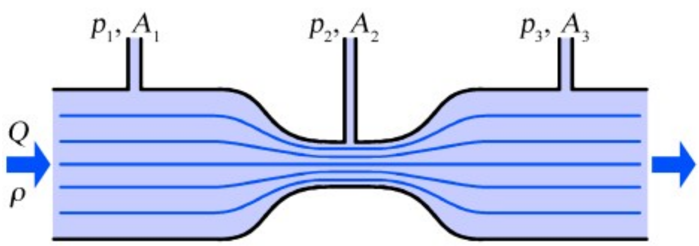
Keywords: pressure sensor; Venturi flow sensor; LTCC; thick-film technology.

1. Introduction

Of over 100 million babies born each year in the western world, 10 million require some form of resuscitation assistance. More than 1 million die from complications of birth asphyxia [1]. Advances in neonatal care have led to significant improvements in the survival of preterm infants, but Chronic Lung Disease (CLD) continues to be a major problem, affecting about 20% of infants who need respiratory assistance [2] and thus being the major long-term pulmonary complication of preterm birth. Neonatal resuscitators used for first-response activities provide a flow of air or air/oxygen mixture to the patient via either a mask or an endotracheal tube. Current neonatal resuscitators monitor and limit delivery pressure and hence avoid the risk of excessive pressure but they do not control the actual tidal air volume exchanged during resuscitation which is important for survival. A new neonatal resuscitator limits delivery pressure and controls tidal volume and respiratory rate simultaneously, requiring accurate measurement of the pressure and the volume flow of air (Fig. 1a) [3]. The target specification for the differential pressure is 50 Pa with an accuracy of 1 Pa to detect the very small flow between 1-4 ml/s necessary for the smallest preterm neonates.



(a)



(b)

Fig.1. (a) neonatal respiratory assistance (source: www.kmedical.co.nz); (b) 3D view & working principle of Venturi tube

used to convert flow into pressure.

To this end, we have developed a flow sensor integrated in the Neonatal Resuscitator/Ventilator using the Venturi principle (Fig. 1b) to convert flow into pressure differences. In this work, after presenting the flow sensor concept, we mainly concentrate on the ultra low pressure sensor: its manufacture and assembly and extensive characterisation.

2. New Resuscitator concept and design

In order to limit delivery pressure and hereby control airway pressure and tidal air volume, the resuscitator makes use of a Venturi tube mounted in parallel to an ultra-low pressure sensor to measure the pressure drop in the Venturi.

2.1. Venturi

The Venturi is characterised by the cross section ratio A_1/A_2 (Fig. 1b), which has a direct impact on the pressure range. Smaller cross sections A_2 of the constriction give higher pressure drops (i.e. increased sensitivity), allowing measurement of smaller flows, but also lead to higher losses. Ideally (not accounting for losses or compressibility), the pressure difference $\Delta p_{ij} = p_i - p_j$ for a volume flow Q of a fluid of density ρ is given by (1):

$$Q^2 = -\frac{2}{\rho} \cdot \frac{p_i - p_j}{A_i^{-2} - A_j^{-2}} \qquad (1)$$

To detect the very low flow close to 1 ml/s, we have used a Venturi tube 37 mm long (Fig 2.). The entry and exit diameters are respectively 15.2 mm and 13 mm. The constriction is 3.5 mm long with a 1.8 mm diameter. The entry and exit angle are 30° and 7°. This Venturi allows the measurement of the pressure range between 0 and 50 Pa with an accuracy requirement of 1 Pa.

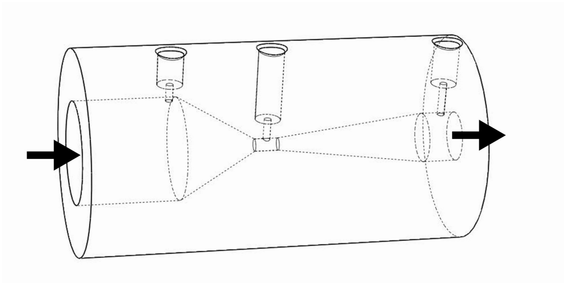


Fig. 2. 3D view of the Venturi tube

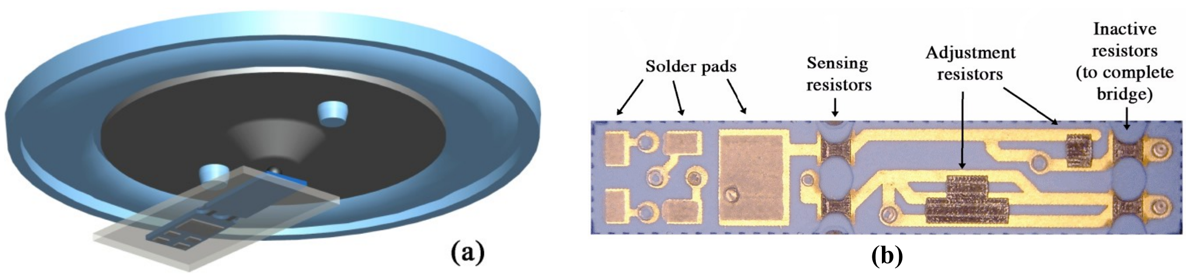
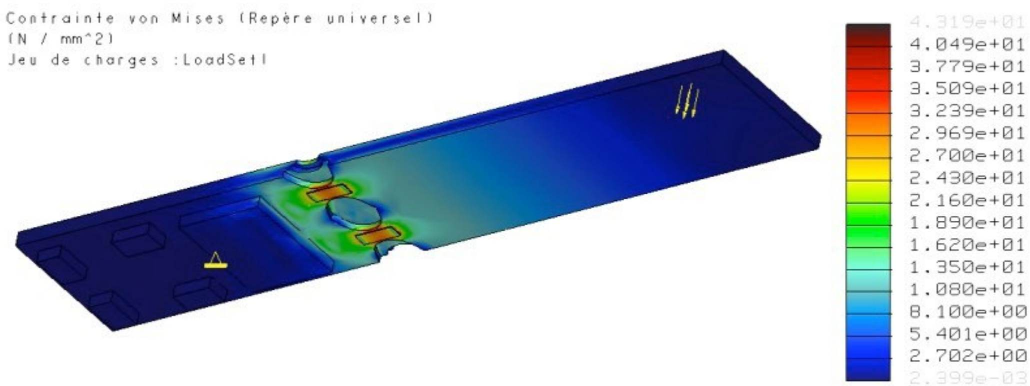


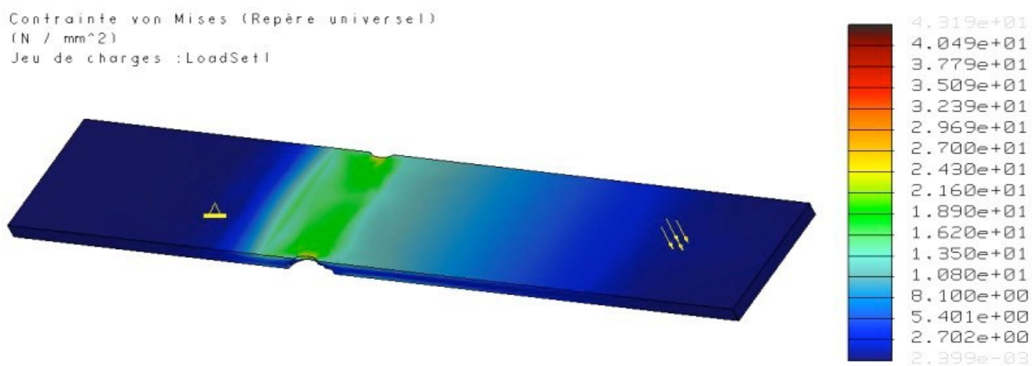
Fig. 3. (a) Elastomer membrane with reinforcing central structure mounted on LTCC cantilever sensor; (b) LTCC cantilever - bottom side.

2.2. Ultra-low pressure sensor

The differential pressure sensor is based on a sensitive membrane-cantilever combination: a very compliant elastomeric membrane with a central rigid area (effective diameter 26 mm, Fig. 3a) efficiently converts the pressure into a force that is measured by the LTCC cantilever force sensor (Fig. 3b). Compared to classical thick-film technology, LTCC allows the fabrication of much more sensitive devices, due to its lower elastic modulus, availability of very thin tapes and high structurability [4,5]. Moreover the LTCC manufacturing process allows cutting out the beam in order to increase the sensitivity of the piezoresistive bridge. To address the challenge of achieving a high signal at low loads, yet with moderate overall deflection, the LTCC design was refined by finite-element modelling (FEM, Fig. 4). Optimisation yielded a two-layer structure: 1) a thick layer (254 / ~210 μm green / fired) carrying the piezoresistors and locally narrowed to concentrate compressive stresses, and 2) a thin opposite layer (114 / 100 μm green / fired) that is not structured. The local narrowing around the resistors concentrates the compressive stresses, optimising sensing performance (Fig. 4a), while the tensile stresses, taken up by the whole width on the other side, remain moderate (Fig. 4b). Another advantage is that the strains are concentrated where they are needed (i.e. for measurement), conserving stiffness. Fabrication of the sensing cantilevers was carried out as in our previous work [5,6], with the DuPont 951 LTCC materials system and DuPont 2041 piezoresistors. The resulting LTCC cantilever was then soldered as an SMD (surface-mount device) component onto a thick-film base that also comprises the amplification and adjustment electronic circuit [6,7]. The resulting force sensor was calibrated for 100 mN nominal force (±1% of the offset and in full scale).



(a)



(b)

Fig. 4. Von Mises stress distribution by FEM on (a) bottom side and (b) top side of cantilever.

3. Characterisation

Seven LTCC force sensors were mounted to the membranes as ultra low pressure sensors and connected to the venturi. Moreover, a commercial relative and differential pressure sensor from Huba Control [0…500 Pa] [8] was also used for comparison. The medical air flow was delivered by a manual brass core oxygen regulator [from 0.5 ml/s] [9] through the venturi. The pressure differences were then measured out from the venturi in parallel by an AirFlow MEDM 500 micromanometer and by the developed sensors as depicted on the Fig. 5. The output-pressure curves of the sensors were then determined (Fig. 6). We can see that the response of five developed sensors give the same sensitivity 0.3 V/Pa and can detect ultra low pressure of 0.2 Pa and very small flows (~1 ml s) through the Venturi. Compared to that, the first pressure detected with the commercial sensor is from 2.3 Pa. As the sensitivities of the LPM force sensors were laser trimmed to be identical within 2%, the observed lower overall sensitivity of the completed pressure sensors (LPM sensors 6-2 and 7-1) can therefore be attributed to assembly problems. Moreover according to the sensitivity of the pressure sensor and the sensitivity of the force sensor, we could calculate the effective radius of the membrane which is 12.4 vs 13.0 mm (cf Fig. 7). The force is distributed through the elastomer bellows but some sensitivity is lost by its finite rigidity. Afterwards, the repeatability of the measurements of the pressure sensors were controlled and exposed on the Fig. 8.

Nevertheless, although Venturis with a narrow constriction are more sensitive (i.e. create a larger pressure drop, see equation 1), the non-recoverable pressure losses (i.e. $\Delta p_{13} = p_1 - p_3$) are similar in magnitude to the measured pressure difference $\Delta p_{12} = p_1 - p_2$. The pressures losses are measured between the inlet and the outlet. With two Huba pressure sensors type 401, we measured the pressure P_{12} and P_{13} in the venturi for flows varying from 10 to 60 ml/s and read the pressure on the sensors. The Fig 9. depicts the sensitivity of the venturi and show the pressures losses. Therefore in a further step, improvements of the design will likely lead to a reduction in the pressure losses to achieve smoother flow transitions and smaller turbulence losses.

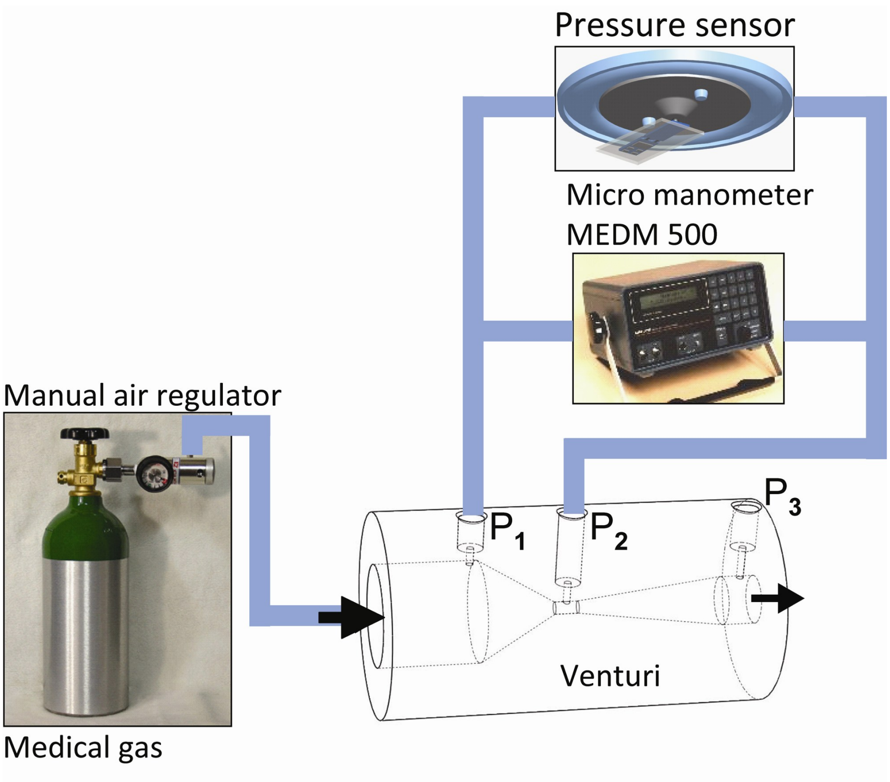
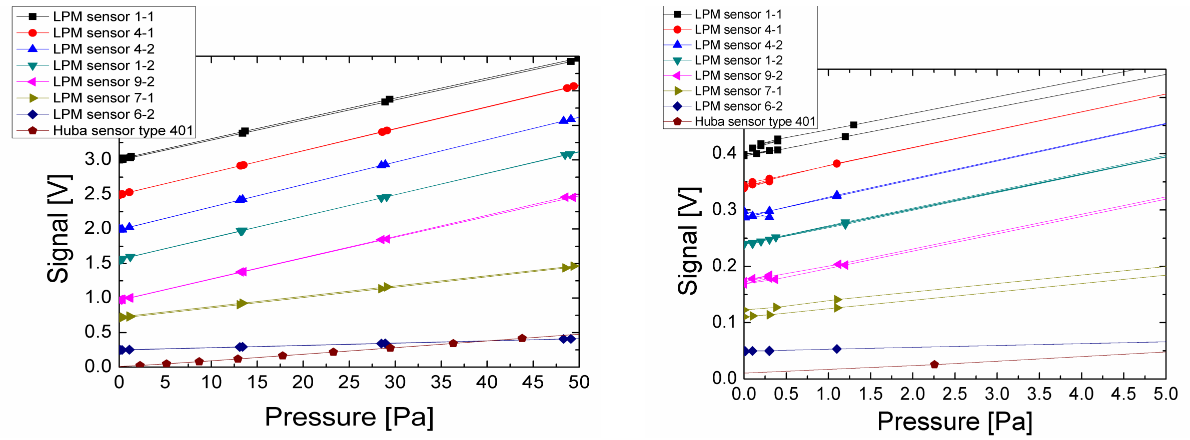


Fig 5. Experimental setup.



(a)

(b)

Fig. 6. Signal of different sensors assembled with the Venturi tube as a function of the pressure (a), with zoom for low flows in (b). Output signals shifted for better readability.

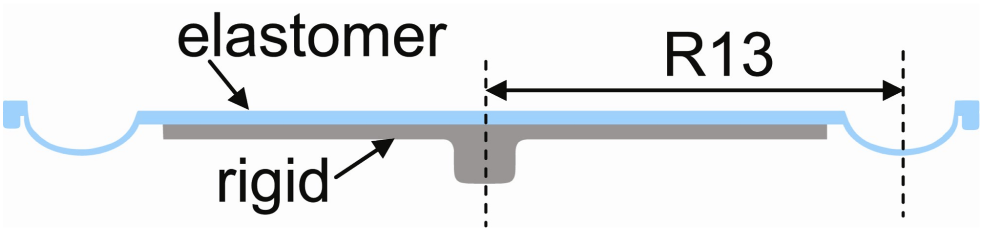


Fig 7. Schematic of the membrane section.

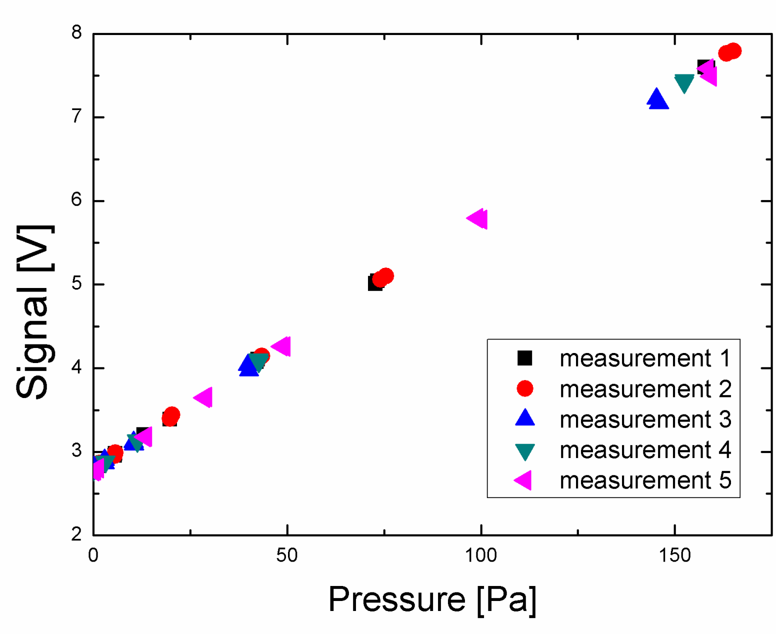


Fig 8. Repeatability of the response of the sensor LPM 9-2 in function of the pressure.

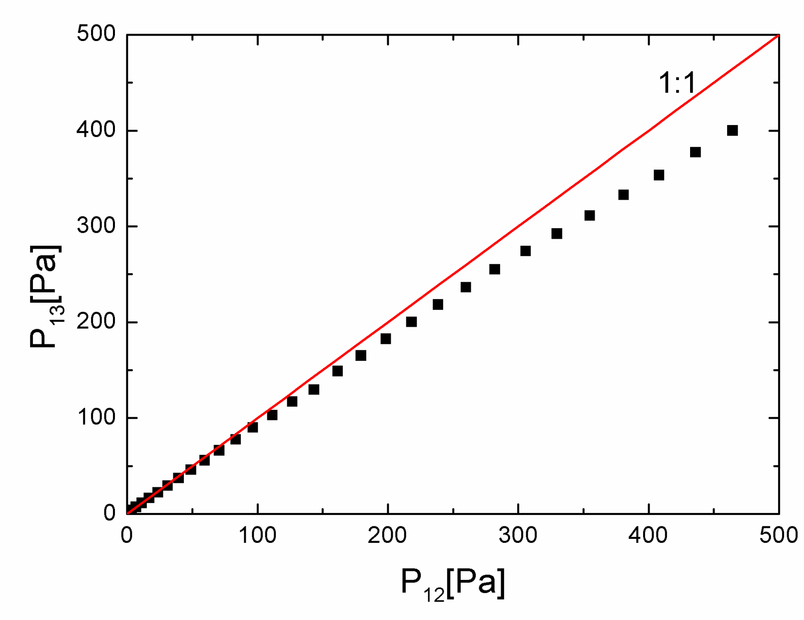


Fig 9. Losses pressures in function of the measured pressure difference

4. Conclusions

We succeeded in developing an ultra-low pressure sensor sensitive to pressure differences under 1 Pa with a sensitivity of 0.3 V/Pa thanks to the clever combination of a very low force sensor and a compliant elastomeric membrane. Assembled to a Venturi, this device allows accurate measurement of very small air flows (~1 ml/s), which is required for new generation Neonatal Resuscitator/Ventilators that control airway pressure, respiratory rate and tidal volume simultaneously. The Venturis used here, however, still need improvement to optimise the generated pressure difference while limiting pressure losses.

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